Load response of an osseointegrated implant used in the treatment of unilateral transfemoral amputation:
An early implant loosening case study

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ABSTRACT

**Background:** Osseointegrated implants for transfemoral amputees facilitate direct load transfer between the prosthetic limb and femur; however, implant loosening is a common complication, and the associated implant-bone loads remain poorly understood. This case study aimed to use patient-specific computational modelling to evaluate bone-implant interface loading during standing and walking in a transfemoral amputee with an osseointegrated implant prior to prosthesis loosening and revision surgery.

**Methods:** One male transfemoral amputee with an osseointegrated implant was recruited (age: 59-yrs, weight: 83kg) and computed tomography (CT) performed on the residual limb approximately 3 months prior to implant failure. Gait analyses were performed, and the CT images used to develop a finite element model of the patient’s implant and surrounding bone. Simulations of static weight bearing, and over-ground walking were then performed.

**Findings:** During standing, maximum and minimum principal strains in trabecular bone adjacent to the implant were 0.26% and -0.30%, respectively. Strains generated at the instant of contralateral toe-off and contralateral heel strike during walking were substantially higher and resulted in local trabecular bone yielding. Specifically, the maximum and minimum principal strains in the thin layer of trabecular bone surrounding the distal end of the implant were 1.15% and -0.98%, respectively.

**Interpretation:** Localised yielding of trabecular bone at the interface between the femur and implant in transfemoral amputee osseointegrated prosthesis recipients may present a risk of implant loosening due to periprosthetic bone fracture during walking.
Rehabilitation exercises should aim to produce implant-bone loading that stimulates bone remodelling to provide effective bone conditioning prior to ambulation.
INTRODUCTION

Osseointegrated implants overcome a number of complications associated with traditional socket-fitted prostheses for the treatment of lower limb amputation, including residual limb pain and skin irritation at residual stump-socket interface, which can occur in up to 62% of recipients (Gallagher Malcolm MacLachlan, Pamela, 2001; Hagberg and Brånemark, 2001; Lyon et al., 2000; Mak et al., 2001). Clinical studies have shown that, compared to socket-fitted prostheses, transfemoral osseointegration reduces tissue damage and residual pain (M Al Muderis et al., 2016; Branemark et al., 2001; Sullivan et al., 2003), while improving patient activity levels, and facilitating larger range of hip joint movement (Brånemark et al., 2014; Hagberg et al., 2008, 2005). In two-year follow-up studies by Brånemark et al., 2014 and Hagberg et al., 2008, implant survival rates were 92% in 51 patients and 94% in 18 patients, respectively. Nebergall et al. (2012) used radiostereometric analysis in 51 patients to show no significant implant migration or rotation at ten-year follow-up. In contrast, a systematic review of outcomes of osseointegrated prostheses for transfemoral amputation reported a maximum revision rate of 67%, with osseointegrated implants loosening in up to 6% of patients, and periprosthetic fracture occurring in up to 9% of cases (van Eck and McGough, 2015). In a study of 51 amputees with osseointegrated implants, mechanical failure of the prosthesis was observed in up to 29% of cases (Brånemark et al., 2019), including damage to the abutment and/or abutment screw that forms the connection between the intramedullary stem and the prosthetic lower limb. Failures have also been reported to the intermedullary stem, with four incidents described across 77 patients (Munjed Al Muderis et al., 2016; Zimel et al., 2016).
Transfemoral amputees with osseointegrated implants typically undergo post-operative rehabilitation for up to six months to encourage integration of bone into the implant prior to ambulation (Hagberg and Branemark, 2009). In the first phase of rehabilitation, a controlled loading regime is applied to incrementally stimulate bone mineralization and strength, whilst minimizing risk of implant overloading. Core and limb strengthening exercises are commonly performed in conjunction with axial weight bearing using a short training prosthesis, and static loads of increasing magnitude gradually applied until the amputee is able to support half their bodyweight without significant pain (Al Muderis et al., 2017b; Hagberg et al., 2005; Hagberg and Branemark, 2009). Some protocols also adopt eccentric prosthesis loading exercises to encourage bone growth in directions other than axial (Hagberg and Branemark, 2009). In the second phase of rehabilitation, prosthetic limbs are fitted, and parallel bars and crutches employed to limit load transmission through the entire prosthesis. The patient may repeat the static weight bearing tasks, and progressively introduce different phases of the gait cycle until they are able to ambulate unaided (Leijendekkers et al., 2017). At present, the loading patterns at the bone-implant interface during rehabilitation tasks such as static weight bearing, and normal walking are poorly understood. As a consequence, over-emphasis on functional rehabilitation tasks that under-load the bone in the residual limb may not adequately prepare the patient for the dynamic loads during walking, and may ultimately result early implant loosening or failure (Huiskes et al., 1987; Turner et al., 1986).

Finite element modeling has been used in several studies to investigate stress-strain distributions and long-term remodeling of bone surrounding osseointegrated
transfemoral implants (Helgason et al., 2009; Lee et al., 2008; Stenlund et al., 2017; Tomaszewski et al., 2010). While these previous studies have been used to describe risk of implant failure, implant loading was achieved using a resultant external force and moment applied directly to the prosthesis. However, non-knee-spanning muscles contribute to the linear and angular acceleration of the knee as well as knee joint-contact loading via dynamic coupling (Zajac et al., 2002); replacing the forces of individual muscles by an equivalent knee load may ultimately underestimate prosthesis force response (Blemker et al., 2007; Herzog et al., 2003). The study by Lee et al., 2008 showed distinctive differences in bone-implant interface stresses between weight-bearing and walking; however, the femur did not incorporate subject-specific bone geometry or material properties, which would have significantly altered the magnitude and distribution of the stress predictions.

The objective of this study was twofold. Firstly, to develop a patient-specific multi-body musculoskeletal model of the residual limb of an amputee with an osseointegrated prosthesis in the period before the implant loosened due to a periprosthetic fracture, and secondly, to use this model to evaluate loading patterns at the bone-implant interface during static weight-bearing with both a short training prosthesis and the entire prosthetic limb, as well as walking with the prosthetic limb. The findings of this case study will help in the development of personalised rehabilitation regimens to improve bone-implant integration and minimize early implant loosening.

MATERIALS AND METHODS

Subject recruitment and imaging
One male subject (age: 59 years, weight: 83 kg, height: 1.90 m) with an above-knee amputation due to an automobile accident was recruited 12 months following primary surgery to implant a transfemoral osseointegrated prosthesis. The implant was an Osseointegrated Prosthetic Limb (OPL; Permedica s.p.a, Milan, Italy) with a cross-screw design (M Al Muderis et al., 2016), connected to a Genium microprocessor controlled knee (Otto Bock 3B1), a pylon with in-built torsion adaptor (Otto Bock 2R21) and dynamic response foot (Otto Bock 1C60, size 28 cm) (Figure 1). The subject had no history of previous surgery, osteoarthritis, or pain in the residual or contralateral limb. This study was approved by the University of Melbourne Human Ethics Committee and written-informed consent was obtained.

The subject’s residual limb was scanned using computed tomography (CT) (Somaton Definition AS+; Siemens Healthineers, Munich, Germany) with a metal artefact reduction sequence and a voxel size of 0.79 x 0.79 x 0.75 mm. The three-dimensional geometry of the femur and osseointegrated implant was digitally reconstructed from the CT images using commercially available software (Mimics version 21.0; Materialise, Leuven, Belgium). Approximately 14 months post-operatively, the osseointegrated implant failed by periprosthetic fracture leading to loosening of the intramedullary stem in the femoral canal, and the implant was revised to a non-cross-screw design OPL. The position of the revised osseointegrated implant within the intramedullary canal was similar to that of the implant used in the initial surgery, as confirmed by aligning the 3D components obtained in the initial surgery with the 2D projection of the revised components from a frontal-plane x-ray (Figure 2). Since the identical prosthetic limbs were used both prior to and following surgery, it is likely that
the positions of the residual limb and hinge axis of the prosthetic knee remained relatively unchanged.

**Motion capture experiments**

Motion analysis experiments were performed with the subject’s revised osseointegrated prosthesis 12 months following the revision surgery. Data collection was performed with the subject standing in a relaxed pose, as well as during over-ground walking at the preferred self-selected speed (Figure 3a). Three-dimensional positions of retro-reflective markers placed on the subject were measured using a 9-camera motion analysis system (Vicon, Oxford Metrics Ltd., UK) sampling at 120 Hz. Ground reaction force data were simultaneously acquired using three instrumented force platforms (AMTI Inc., Watertown, MA) sampling at 1080 Hz. Marker trajectories were filtered using a low-pass, fourth-order Butterworth filter with a cut-off frequency of 10 Hz, while ground reaction force was filtered at 60 Hz using a four-order Butterworth filter.

**Computational modeling**

A rigid-body musculoskeletal model of the amputee’s residual limb prior to prosthesis failure was developed using a subject-specific computational modelling framework (Figure 3b,c). The 12-segment, 23-degree-of-freedom rigid-body model actuated by 76 Hill-type muscle-tendon units was created by scaling a generic musculoskeletal model of an adult to the mass and anthropometry of the patient (Ackland et al., 2012). The 3D geometry of the patient’s residual femur was imported into the model, and the distal femur, shank and foot segments replaced with the osseointegrated implant and lower limb. The knee joint was modelled as an ideal hinge, with the external torque of the microprocessor-controlled knee represented by a coordinate (reserve)
actuator at the knee. Intact femoral muscles were attached to the residual femur according
to the centre of their tendon attachments which were located relative to bony landmarks
on the femur using the 3D anatomical atlas software Complete Anatomy (version 5.1.1;
3D4Medical, Dublin, Ireland). Surgically reconstructed muscles were re-attached to the
distal end of the femur using a myodesis technique and any absent or dysfunctional
musculature in the residual limb were removed from the model.

The mass and inertial properties of the residual limb segments was determined by
representing the associated components using geometric shapes of uniform density. The
residual limb above the knee was represented as the frustum of a right circular cone,
where the maximum and minimum circumference and height were measured with a
measuring tape and a uniform tissue density was assumed of 1.1 g/cm$^3$ (Smith et al.,
2014). The mass, centre-of-mass and mass-moment of inertia components were computed
for this geometry using CAD software (Solidworks, Dassault Systems, Paris, France).
The combined mass of the knee and pylon segments was measured using digital scales
and represented as a solid cylinder with dimensions equal to their maximum
circumference and total height, as evaluated using measuring tape. The combined knee
and pylon was assigned a constant density for carbon fibre material (1.65 g/cm$^3$), and the
referred to by the researchers as the ‘centre of mass’ was set at the centre of the cylinder with the mass moment
of inertia calculated using standard engineering formula (Eqs 1 and 2). The foot segment
was modelled as a triangular prism with its height and width determined using the tape
measure. Assuming a uniform density of a carbon fibre material, the centre of mass and
mass moments of inertia were computed using the CAD software. The position of the
hinge axis on the prosthetic knee was determined by placing two reflective markers
coincident with the rotation axis on the prosthetic limb. The correct positioning of these markers on the axis was verified by examining their 3D measurements recorded by the motion capture system during flexion of the prosthetic limb.

\[ I_x = I_y = \frac{1}{12} m(3r^2 + h^2) \]  
\[ I_z = \frac{1}{2} mr^2 \]

where \( I \) is the mass moment of inertia, the subscripts \( x \) and \( y \) denote the radial directions and \( z \) denotes the axial direction, \( m \) is the mass, \( r \) is the radius, \( h \) is the cylinder height.

The musculoskeletal model was used to simulate static weight bearing conditions with a short training prosthesis as well as the prosthetic limb. Simulations of walking were also performed, specifically, four gait events during the stance phase that included ipsilateral heel strike, contralateral toe-off, contralateral heel strike and ipsilateral toe-off (Figure 4). For each simulation, inverse kinematics was applied to the motion capture data. Net internal joint moments were then calculated using inverse dynamics and decomposed into muscle forces using static optimization (Dorn et al., 2012). For static weight bearing with the short training prosthesis, 50% body-weight was applied to both the abutment of the intramedullary stem and to the foot of the intact limb.

A deformable model of the residual limb and implant was developed in a commercial finite element modelling package (Abaqus 2017; Simulia, RI, USA) (Figure 3D). The model was created and output data expressed in the femoral coordinate system, which was located at the centre of the femoral head with the \( y \)-axis pointing cranially from the prosthetic knee hinge axis centre to the femoral head centre, the \( x \)-axis pointing anteriorly and perpendicular to the plane defined by the \( y \)-axis and the prosthetic knee.
hinge axis, and the z-axis calculated as the cross-product of the x- and y-axes. The three-dimensional geometries of the femoral cortical bone, trabecular bone and implant were meshed using tetrahedron elements using Hypermesh (version 2017.2; Altair Engineering, Michigan, USA).

To prevent a stress concentration at the connection between the intramedullary stem and cross-screw, a 0.5 mm fillet was added to the outer edge of the hole on the cross-screw and the mesh density was increased at the interface between the components. The average element size and total number of elements of each component are provided in Table 1. The intramedullary stem and cross-screw were rigidly tied to one another, and the thread on the screw modelled as a uniform 10 mm diameter cylinder. The implant and bone were modelled as linear-elastic solids, with material properties for the implant set to a Ti6Al4V titanium alloy (Al Muderis et al., 2017a), whilst the bone was assigned a heterogeneous apparent bone mineral density ($\rho_{\text{app}}$) with an elastic modulus derived from the CT data (Table 1). The intramedullary stem and the threaded region of the cross-screw were assumed to be completely osseointegrated with the surrounding bone, and these surfaces were tied to prevent relative motion between respective nodes at the contact interfaces. The surface of the femoral head was clamped three-dimensionally to model contact against the acetabulum. This boundary condition was chosen because the resultant hip joint contact force was approximately at the centre of the femoral head, which ensured approximately equivalent boundary condition positions, joint forces and moments with that of the musculoskeletal model. Although this clamped boundary condition would have differed from joint contact in a native joint, this difference was
assumed to have only a localised contribution to the femoral stress predictions in the vicinity of the femoral head.

For simulations of standing (training prosthesis and entire prosthetic limb) and walking, the rigid-body musculoskeletal model predictions of kinematics, inertial forces, muscle forces and knee joint reaction forces were applied to the finite element model. Twenty-seven muscle forces were included in the model, with each applied across several nodes on the bone in the vicinity of the muscle attachment. The knee-joint reaction force and moment were transmitted to the distal end of the intramedullary stem via a rigid multi-point constraint (MPC). The inertial forces and moments due to linear and angular femoral acceleration were applied at the distal end of the intramedullary stem. Peak von Mises stresses, and maximum and minimum principal strains were calculated for the implant, and femoral trabecular bone and cortical bone in four discrete femoral regions: the femoral neck, intertrochanteric, subtrochanteric and diaphysis (Figure 5). A strain-based failure criterion was used to identify trabecular bone yielding, whereas a stress-based criterion was used to identify cortical yielding/fracture and implant yielding (Table 1). To verify model consistency, the hip joint reaction force calculated using the musculoskeletal model was compared to the hip joint-contact force estimated using the finite element model. To evaluate sensitivity of the results to changes in mesh density, the entire FE model mesh density was increased to the highest amount possible (approximately 3.60 million elements) and used to simulate the contralateral toe-off load case (see Supplementary Material). Compared to the nominal mesh density, the stress and strain predictions had an average difference less than 10% for each component, which
was considered sufficient to provide reasonable comparisons between the load cases considered in the current study.

**RESULTS**

The largest muscle contribution to support the residual limb during standing was from adductor brevis, adductor longus, gluteus medius (anterior) and iliacus, with each generating between 62.1 and 248.9 N (Table 2). In comparison, there were greater forces generated by particular hip-spanning muscles during the stance phase of walking, with peak forces greater than 500 N for gluteus maximus, gluteus medius, rectus femoris, iliacus and psoas (Table 2). The resultant force at the abutment of the training prosthesis and knee joint of the lower limb prosthesis were similar during standing (406.7 N and 375.6 N, respectively) (Table 2), and each was less than half of the peak knee force during the stance phase of walking (894.8 N at contralateral toe-off). While the training prosthesis produced no external knee joint moment during standing, the resultant knee joint moment for the lower limb prosthesis was 46.2% of the peak observed during stance (41.0 Nm). The maximum linear and angular accelerations calculated for the residual limb centre of mass during the stance phase of walking was 7.6 m/s² and 49.1 rad/s², respectively (Table 2). The hip joint contact forces predicted by the finite element model were within 5% of those predicted by the rigid-body musculoskeletal model (Table 3).

The peak von Mises stress of the intramedullary stem when standing with the training prosthesis and lower limb prosthesis occurred superiorly along the medial edge of its connection with the cross-screw (132.7 MPa and 140.5 MPa, respectively) (Table 4) (Figure 6). In contrast, the peak stress on the intramedullary stem increased from 39.8 MPa at ipsilateral heel strike to 522.8 MPa at contralateral toe-off, before decreasing to
25.6 MPa at ipsilateral toe-off. The peak stresses at the cross-screw when standing with
the training or lower limb prosthesis (126.8 MPa and 131.0 MPa, respectively), were
much lower than the peak stress incurred during stance in this region (501.8 MPa at
contralateral toe-off).

For femoral cortical bone, the peak von Mises stress in each region of the femur
ranged from 6.1 MPa to 23.2 MPa for the training prosthesis and 10.3 MPa to 41.1 MPa
for the lower limb prosthesis during standing (Table 4). During stance, the
intertrochanteric region and diaphysis had a peak cortical stress at contralateral toe-off of
98.0 MPa and 78.4 MPa, respectively, whilst the femoral neck and subtrochanteric region
had a peak stress at contralateral heel-strike of 74.8 MPa and 82.6 MPa, respectively.
During walking and standing, the stresses in femoral cortical bone were lower than those
of the yield strength of cortical bone (Table 1). The peak stresses in the cortical bone
occurred in the vicinity of muscle attachments on the greater and lesser trochanter and at
the distal end of the diaphysis, although large stresses were also apparent in the femoral
neck for contralateral toe-off and contralateral heel strike (Figure 7).

Considering all regions of the femur during standing with the training prosthesis,
the trabecular bone had a maximum and minimum principal strain of 0.25% and -0.20%,
respectively (Table 4). These strains were slightly greater for the load case with the lower
limb prosthesis (0.54% and -0.38%, respectively). During walking, a maximum principal
strain of 1.15% was observed in the diaphysis at contralateral heel-strike, while a
minimum principal strain of -0.80% occurred in the diaphysis at contralateral toe-off.
Strains at or slightly below these maxima were observed in the thin layer of trabecular
bone that was present at the interface the implant (Table 4), specifically at the most distal
region of the femur (Figure 8). The peak strains during standing did not result in trabecular bone damage, whereas there were localised regions of bone yielding observed during the stance phase of walking (Table 1) (Figure 8).

**DISCUSSION**

The objective of this case study was to employ patient-specific multi-body musculoskeletal modelling to evaluate the load response of an osseointegrated transfemoral implant prior to its failure due to loosening. This was achieved by quantifying forces, stresses and strains generated during static weight bearing with a training prosthesis and full lower limb prosthesis, as well as during over-ground walking with the lower limb prosthesis. While post-operative static-weight bearing tasks are performed to condition femoral bone prior to ambulation, there is paucity of quantitative data describing resultant loads at the implant-bone interface in osseointegrated transfemoral prostheses. In the present study, the predicted loads during walking may have contributed to failure of the implant-bone fixation after surgery. The findings emphasise the importance of correct prosthesis selection and intraoperative placement, as well as rigorous post-operative rehabilitation prior to ambulation.

The osseointegrated implant experienced the largest stresses at the boundary between the intermedullary stem and screw, and the abutment (Figure 6). This finding may be due to the presence of a prominent bending moment, since the length of the stem and diaphysis provide a considerable moment arm of the distally-located forces at knee joint force. Although the stresses at these locations were below the yielding threshold, it is nonetheless suggested that future osseointegrated implant designs mitigate these dynamic bending-generated loads, for example, using alternative implant geometry such
as support ribs or radiused edges, or by using less-stiff biomaterials such as 3D printed lattice structures (Chen et al., 2016). Although the locations of the observed peak stresses in the cortical bone tended to occur at the muscle attachments (Figure 7), this result was likely due to the small attachment areas assumed in the finite element model (Polgar et al., 2003). The small attachment regions were applied to achieve the same moment arms of the muscles in the musculoskeletal model which act as point loads. Despite this result, the relatively large stresses observed in the femoral neck during contralateral toe-off and contralateral heel-strike are consistent with studies examining peak stresses in the proximal femur during stance (Martelli et al., 2014; Van Rietbergen et al., 2003).

No previous studies have described failures related to the cross-screw, which was the location of the greatest implant stresses in the current study. However, this design is only available in the OPL implant system and is applied to short femurs less than 16 cm in length (Al Muderis et al., 2018), and thus a limited number of individuals have received this implant which inherently restricts the number of patients that would experience potential implant failure. Further, mechanical failures of implanted osseointegrated components without a cross-screw are relatively rare, with only four cases on intramedullary stem failure reported across 77 patients in two studies (M Al Muderis et al., 2016; Zimel et al., 2016). These studies either did not specify the implant failure location or used a considerably different implant design, which prevents a direct comparison to the peak implant stress locations in the current study. Mechanical failure has been reported in the OPRA implant system at the abutment and/or abutment screw that connects to the intramedullary stem and extends out from the bone and skin (Brånemark et al., 2019). Indeed, the peak stress in the intermedullary stem was observed
in a similar distal location, external to the bone (163.8 MPa at contralateral toe-off; Table 4), which is in close agreement with the stress distributions predicted in the current study. During the single-support phase of stance, the maximum and minimum principal strains for the trabecular bone were sufficient to induce yielding in the femoral neck, intertrochanteric and diaphysis regions (Table 4). In most cases, these were localized to small regions at the interface between the bone and the implant, especially for trabecular bone that appeared to have formed at the distal end of the diaphysis (Figure 8). Localised yielding or microdamage may initiate bone remodeling that potentially strengthens the bone in the vicinity of the implant (Burr et al., 1997; Leucht et al., 2007; Nagaraja et al., 2005). Conversely, excessive yielding may present a risk of periprosthetic bone fracture, or implant migration that may lead to aseptic loosening (Taylor and Prendergast, 2015). The predicted bone yielding sites at the proximal and distal regions of the femur compare well with reported fracture locations associated with osseointegrated implantation where femoral fractures occurred proximal from the implant (Al Muderis et al., 2018; Munjed Al Muderis et al., 2016; Juhnke et al., 2015), or periprosthetic fractures occurred in the proximal or distal region of bone surrounding the implant (McGough et al., 2017; Tyler et al., 2009; Zimel et al., 2016). It is not possible to compare the yielding sites with cases of aseptic loosening, as the initial site of implant loosening has not been reported and cannot easily be determined retrospectively. A comparison between the predicted yielding sites and implant loosening would be well suited to mechanical testing on cadaveric specimens using dynamic imaging techniques such as dual plane fluoroscopy (Tsai et al., 2013), and is recommended in future work.
In the present study it was observed that femoral stresses and strains for each of
the static weight bearing conditions were much lower than those during walking (Figure
8) and were below the thresholds necessary to induce trabecular or cortical bone yielding.
Therefore, the standing-based rehabilitation task investigated may be considered
inadequate to stimulate bone remodelling to sufficiently condition bone prior to walking.
Pre-rehabilitation strategies ought to be developed to gradually introduce bone strains
closer to those during walking to promote more extensive bone remodelling, for example
sit-to-stand, step-up or squat tasks (Martelli et al., 2014).

The results presented in this study are in reasonable agreement with previous
finite element analyses of transfemoral osseointegrated implants. The peak stress
observed at the distal end of the intermedullary stem at contralateral toe-off (161.7 MPa;
Figure 5), was in reasonable agreement with the reported a peak stress of a transfemoral
osseointegrated implant of 137 MPa at its distal end during midstance (Xu et al., 2006).
Another finite element modeling study of an osseointegrated implant found that the peak
von Mises stress of bone surrounding the implant was 8.5 MPa, occurring at late stance
(Lee et al., 2008), which is similar to the peak stress observed for this region in the
current study of 10.6 MPa at contralateral heel strike. Finally, a previous modelling study
predicted a peak von Mises equivalent strain in the diaphysis of an osseointegrated
implant of 0.18% (Xu et al., 2016). This was of a similar magnitude to our peak strains
predicted in the diaphysis for the training prosthesis (maximum, minimum principal
strain: 0.25%, -0.14%) and full lower limb prosthesis during standing (maximum,
minimum principal strain: 0.28%, -0.20%). While the peak strains predicted in the current
study for stance were much greater, this may be due to the large bending moment
generated by the microprocessor-controlled knee (approximately 40 Nm at contralateral toe-off), which was not modeled in the study of Xu et al. (2016).

There are limitations of this study that ought to be considered. First, it is unclear whether the regions of low bone density at the bone-implant interface represented the true bone properties or were the consequence of CT artifacts from the metallic implant such as beam hardening, photon starvation or scatter (Pessis et al., 2013). However, a metal artefact reduction algorithm was used to minimise this undesirable effect, though artefact may still affect the CT measurements of bone density close to the implant (Koff et al., 2017; Wellenberg et al., 2018). Second, the tied contact between the implant and bone simulated the case of complete osseointegration of the implant and did not consider the possibility of micro-motion at the bone-implant interface. Third, the thread on the implant was not modeled to avoid stress discontinuities, which may have affected the stress distribution in the vicinity of the bone-implant interface. Finally, due to radiation dosage, a CT scan was not performed following the implant revision surgery; however, the revised implant position was assessed using a frontal-plane x-ray. Comparing the 3D alignment of the initial implant with the 2D projection of the revised implant indicated that the alignment was similar (Figure 2), suggesting the revised implant was press-fit into the intramedullary canal and that its position was not substantially altered in the revision surgery.

The present study investigated loading on a transfemoral osseointegrated prosthesis that, after 14 months post-operatively, failed by early loosening and was subsequently revised. By using multi-body musculoskeletal modelling, it was shown that strains at the prosthesis-bone interface during standing were substantially lower than
those generated during walking. Localised regions of yielding of trabecular bone in the vicinity of the distal femur were observed during walking, which may ultimately present risk of periprosthetic fracture or implant migration, particularly in cases of insufficient osseointegration and over-emphasis on static weight bearing prior to ambulation. Personalised rehabilitation protocols that target unassisted ambulation ought to involve dynamic tasks that transition the osseointegrated limb between low intensity static weight-bearing and those associated with walking.
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TABLES

Table 1: Material properties, average element size and number of elements used in the finite element model of the transfemoral amputee’s residual limb, including osseointegrated prosthesis and femoral cortical and trabecular bone.

<table>
<thead>
<tr>
<th>Component</th>
<th>Material</th>
<th>Density (g/cm³)</th>
<th>Elastic modulus (GPa)</th>
<th>Poisson’s ratio</th>
<th>Yield limit</th>
<th>Ultimate strength</th>
<th>Average element size (mm)</th>
<th>Number of finite elements</th>
</tr>
</thead>
<tbody>
<tr>
<td>Intramedullary stem</td>
<td>Ti6Al4V</td>
<td>4.4</td>
<td>115</td>
<td>0.33</td>
<td>σ=795 MPa</td>
<td>σ=860 MPa</td>
<td>0.37</td>
<td>715,217</td>
</tr>
<tr>
<td>Cross-screw</td>
<td>Ti6Al4V</td>
<td>4.4</td>
<td>115</td>
<td>0.33</td>
<td>σ=795 MPa</td>
<td>σ=860 MPa</td>
<td>0.23</td>
<td>544,653</td>
</tr>
<tr>
<td>Trabecular bone</td>
<td>0.0-1.0</td>
<td>0.5-5.7</td>
<td>0.3</td>
<td></td>
<td>εY,comp = 0.70%</td>
<td>N/A</td>
<td>0.85</td>
<td>587,105</td>
</tr>
<tr>
<td>Femur</td>
<td>1.0-2.0</td>
<td>5.7-19.7</td>
<td>0.3</td>
<td></td>
<td>εY,tens = 0.61%</td>
<td>N/A</td>
<td>0.84</td>
<td>119,910</td>
</tr>
<tr>
<td>Cortical bone</td>
<td>1.0-2.0</td>
<td>5.7-19.7</td>
<td>0.3</td>
<td></td>
<td>σ=121.0 MPa</td>
<td>σ=140.0 MPa</td>
<td>0.84</td>
<td>119,910</td>
</tr>
</tbody>
</table>

Ti6Al4V properties from (Oldani and Dominguez, 2012)

Bone densities corresponds to apparent bone mineral density (ρ_{app}) scaled linearly from CT grayscale values to the normal range of femur (Morgan et al., 2003). The threshold for ρ_{app} between trabecular and cortical bone was set as 1.0 g/cm³. Elastic modulus was calculated from apparent bone mineral density using E=6850ρ_{app}^{1.49} (Morgan et al., 2003). The yield strains for trabecular and bone were previously published (Morgan and Keaveny, 2001), as were the yield and ultimate stress for cortical bone (Carter et al., 1981).
Table 2: Magnitudes of the muscle forces (N) calculated using the patient-specific musculoskeletal model and used in the finite element model simulations of over-ground walking and a static standing posture.

<table>
<thead>
<tr>
<th>Component</th>
<th>Quantity</th>
<th>Gait event</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Standing: Training prosthesis</td>
<td>Standing: Full lower limb prosthesis</td>
</tr>
<tr>
<td>Adductor Brevis</td>
<td>70.5</td>
<td>102.5</td>
</tr>
<tr>
<td>Adductor Longus</td>
<td>167.7</td>
<td>248.9</td>
</tr>
<tr>
<td>Adductor Magnus (Superior)</td>
<td>26.9</td>
<td>39.3</td>
</tr>
<tr>
<td>Adductor Magnus (Medial)</td>
<td>9.1</td>
<td>12.4</td>
</tr>
<tr>
<td>Adductor Magnus (Inferior)</td>
<td>37.7</td>
<td>48.2</td>
</tr>
<tr>
<td>Biceps Femoris Long Head</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>Gemellus</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>Gluteus Maximus (Superior)</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>Gluteus Maximus (Medial)</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>Gluteus Maximus (Inferior)</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>Gluteus Medius (Anterior)</td>
<td>99.7</td>
<td>101.3</td>
</tr>
<tr>
<td>Gluteus Medius (Medial)</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>Gluteus Medius (Posterior)</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>Gluteus Minimus (Anterior)</td>
<td>3.4</td>
<td>3.4</td>
</tr>
<tr>
<td>Gluteus Minimus (Medial)</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>Gluteus Minimus (Posterior)</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>Gracilis</td>
<td>4.2</td>
<td>5.9</td>
</tr>
<tr>
<td>Iliacus</td>
<td>62.1</td>
<td>122.1</td>
</tr>
<tr>
<td>Pectineus</td>
<td>13.3</td>
<td>20.0</td>
</tr>
<tr>
<td>Piriformis</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>Muscles</td>
<td>Psoas</td>
<td>Quadriceps Femoris</td>
</tr>
<tr>
<td>-------------------------</td>
<td>-------</td>
<td>------------------</td>
</tr>
<tr>
<td></td>
<td>22.7</td>
<td>0.0</td>
</tr>
<tr>
<td></td>
<td>73.7</td>
<td>0.0</td>
</tr>
<tr>
<td></td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td></td>
<td>0.0</td>
<td>0.0</td>
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<tr>
<td></td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td></td>
<td>0.0</td>
<td>0.0</td>
</tr>
</tbody>
</table>

* Load application point for training condition was at the distal end of the abutment rather than the knee joint.
Table 3: Hip joint reaction force predicted by rigid body and finite element models for training, standing and during the stance phase of walking. The force components are provided in the local hip coordinate system*.

<table>
<thead>
<tr>
<th></th>
<th>Standing: Training prosthesis</th>
<th>Standing: Full lower limb prosthesis</th>
<th>Ipsilateral heel strike</th>
<th>Contralateral toe-off</th>
<th>Contralateral heel strike</th>
<th>Ipsilateral toe-off</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Rigid-body</td>
<td>Finite element</td>
<td>Rigid-body</td>
<td>Finite element</td>
<td>Rigid-body</td>
<td>Finite element</td>
</tr>
<tr>
<td><strong>Fx (N)</strong></td>
<td>-192.0</td>
<td>-191.8</td>
<td>-331.7</td>
<td>-331.5</td>
<td>-76.5</td>
<td>-74.5</td>
</tr>
<tr>
<td><strong>Fy (N)</strong></td>
<td>-843.0</td>
<td>-843.4</td>
<td>-992.7</td>
<td>-992.7</td>
<td>-338.1</td>
<td>-340.4</td>
</tr>
<tr>
<td><strong>Fz (N)</strong></td>
<td>187.8</td>
<td>188.1</td>
<td>270.5</td>
<td>269.0</td>
<td>156.8</td>
<td>156.4</td>
</tr>
</tbody>
</table>

* The positive x-direction, y-direction and z-direction is anterior, superior and lateral, respectively.
Table 4: Maximum von Mises stress (MPa) for the osseointegrated prosthesis, and maximum and minimum principal strains for the femoral cortical and trabecular bone. Data are given for standing as well as the stance phase of gait, including heel strike, contralateral toe-off, contralateral heel strike, and toe-off.

<table>
<thead>
<tr>
<th>Section</th>
<th>Variable</th>
<th>Component</th>
<th>Standing: Training prosthesis</th>
<th>Standing: Full lower limb prosthesis</th>
<th>Ipsilateral heel strike</th>
<th>Contralateral toe-off</th>
<th>Contralateral heel strike</th>
<th>Ipsilateral toe-off</th>
</tr>
</thead>
<tbody>
<tr>
<td>Implant</td>
<td>von Mises stress (MPa)</td>
<td>Intramedullary stem</td>
<td>132.7</td>
<td>140.5</td>
<td>39.8</td>
<td>522.8</td>
<td>475.4</td>
<td>25.6</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Cross-screw</td>
<td>126.8</td>
<td>131.0</td>
<td>39.9</td>
<td>501.8</td>
<td>451.6</td>
<td>11.2</td>
</tr>
<tr>
<td>Cortical bone</td>
<td></td>
<td>Femoral neck</td>
<td>12.0</td>
<td>13.7</td>
<td>5.4</td>
<td>52.5</td>
<td>74.8</td>
<td>1.6</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Intertrochanteric</td>
<td>18.8</td>
<td>41.1</td>
<td>17.0</td>
<td>98.0</td>
<td>51.4</td>
<td>3.3</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Subtrochanteric</td>
<td>6.1</td>
<td>10.3</td>
<td>4.8</td>
<td>38.2</td>
<td>82.6</td>
<td>4.8</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Diaphysis</td>
<td>23.2</td>
<td>34.7</td>
<td>3.5</td>
<td>78.4</td>
<td>26.0</td>
<td>13.9</td>
</tr>
<tr>
<td>Trabecular bone</td>
<td>Max, Min Principal Strain (%)</td>
<td>All regions of femur</td>
<td>0.22,-0.20</td>
<td>0.26,-0.21</td>
<td>0.12,-0.12</td>
<td>0.84,-0.80</td>
<td>1.15,-0.74</td>
<td>0.20,-0.12</td>
</tr>
<tr>
<td>surrounding implant</td>
<td></td>
<td>Femoral neck</td>
<td>0.12,-0.20</td>
<td>0.14,-0.21</td>
<td>0.06,-0.08</td>
<td>0.52,-0.73</td>
<td>0.41,-0.61</td>
<td>0.03,-0.05</td>
</tr>
<tr>
<td>Trabecular bone</td>
<td>Max, Min Principal Strain (%)</td>
<td>Intertrochanteric</td>
<td>0.25,-0.17</td>
<td>0.54,-0.38</td>
<td>0.11,-0.13</td>
<td>0.75,-0.79</td>
<td>0.46,-0.62</td>
<td>0.03,-0.03</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Subtrochanteric</td>
<td>0.04,-0.06</td>
<td>0.05,-0.08</td>
<td>0.03,-0.03</td>
<td>0.21,-0.27</td>
<td>0.40,-0.32</td>
<td>0.07,0.02</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Diaphysis</td>
<td>0.24,-0.14</td>
<td>0.28,-0.20</td>
<td>0.13,-0.09</td>
<td>0.84,-0.80</td>
<td>1.15,-0.74</td>
<td>0.20,-0.14</td>
</tr>
</tbody>
</table>
FIGURE CAPTIONS

Figure 1: Osseointegrated transfemoral implant modelled in the present study prior to failure by early loosening. The specific regions of the intramedullary stem and cross-screw are indicated, as well as their implanted position in the femur.

Figure 2: Comparison of relative positions of initial osseointegrated implant and femur obtained using CT to the positions of the revised osseointegrated implant and femur obtained using frontal plane X-ray including frontal plane X-ray following the revision surgery, with the 3D geometry of the initial implant and femur aligned to the outline of the femur (a) frontal plane x-ray with segmented outlines indicating the femur (purple), revised implant (green) and the cavity from the removed cross-screw (blue), (b) 3D geometry of initial implant and femur (c), and 2D geometry of femur, revised implant (intramedullary stem) and the cavity from the removed cross-screw (d).

Figure 3: Framework for developing personalised computational model of the residual limb of a transfemoral amputee including over-ground walking experiments performed in a human movement laboratory (a), rigid-body musculoskeletal model of amputee (b), and details of the transfemoral osseointegrated implant, including the intramedullary stem and cross-screw (c), and finite element model of osseointegrated implant and femur (d). Muscle forces are indicated by red arrows. The multi-point constraint (MPC) connection indicated by the blue lines was used to transmit the
knee reaction force ($F_{\text{knee}}$) and moment ($M_{\text{knee}}$) to the prosthesis abutment. The muscles rectus femoris, sartorius and tensor fasciae latae are omitted for visualization purposes.

**Figure 4:** Tasks modelled in the present study including standing using a training prosthesis (a), and full lower limb prosthesis (b). Also investigated was the stance phase of gait with the complete lower limb prosthesis including ipsilateral heel strike (c) contralateral toe-off (d) contralateral heel strike (e), and ipsilateral toe-off (f).

**Figure 5:** Regions of the femur used in the finite element model analysis. These included the femoral neck, intertrochanteric, subtrochanteric and diaphysis.

**Figure 6:** Von-Mises stress distribution in the osseointegrated implant for each load case. The location of the peak stress for the stem and screw are reported for standing using a training prosthesis (a), and full lower limb prosthesis (b). Stress distributions were also investigated for the stance phase of gait with the complete lower limb prosthesis including ipsilateral heel strike (c) contralateral toe-off (d) contralateral heel strike (e), and ipsilateral toe-off (f).

**Figure 7:** Von-Mises stress distribution in the femoral cortical bone for each load case. The location of the peak stress for the entire femur is indicated for standing using a training prosthesis (a), and full lower limb prosthesis (b). Stress distributions were also investigated for the stance phase of gait with the complete lower limb prosthesis including ipsilateral heel strike (c)
contralateral toe-off (d) contralateral heel strike (e), and ipsilateral toe-off (f).

Figure 8: The maximum and minimum principal strain (top row and bottom row, respectively) in the trabecular bone surrounding the implant. Regions of bone exceeding the trabecular yield strength are shown in grey. Data are given for standing using a training prosthesis (a), and full lower limb prosthesis (b). Data are also provided for the stance phase of gait with the complete lower limb prosthesis including ipsilateral heel strike (c) contralateral toe-off (d) contralateral heel strike (e), and ipsilateral toe-off (f).
Figure 1

- Femur
- Shaft
- Tapered flange
- Abutment
- Intramedullary stem
- Prosthesis assembly
- Thread
- Femoral neck screw
- Rod
Figure 2
Figure 4
Figure 6

(a) Max: 132.7 MPa
(b) Max: 140.5 MPa
(c) Max: 39.9 MPa

(d) Max: 522.8 MPa
(e) Max: 475.4 MPa
(f) Max: 25.6 MPa
Figure 7

(a) Max: 41.1 MPa
(b) Max: 23.2 MPa
(c) Max: 17.0 MPa
(d) Max: 98.0 MPa
(e) Max: 82.6 MPa
(f) Max: 13.9 MPa